prevent toe drag. A Safe State can be used to shut off the device when any unexpected circumstances occur. The triggers or transitional parameters for the finite state machine are shown in **FIG. 3**.

[**0031**] For a gait cycle in accordance with one embodiment, Contact **1** begins when the foot switch within the heel was compressed. In this embodiment, the transition into Contact **2** occurred when the Ground Reaction Force (GRF), equal to the sum of all six force transducers, was greater than On Ground, equal to about 60 N, and when the ankle was in dorsiflexion. The ankle was considered to be in dorsiflexion when the angle between the tibia and foot was less than 90°. In this embodiment, On Ground was set to about 60 N because this particular value reliably discerned ground contact from noise during swing. Contact **2** ended when the GRF was less than On Ground. In fact, the transition into Swing always occurred when the GRF was less than On Ground. The controller transitioned to the Safe State when any of the force or angle sensory signals went beyond a specified normal operating range. In this embodiment, the range for each force sensor was about 1000 N, the maximum force that any one sensor should measure in walking for a 90 kg person. The acceptable range for the angle sensor was about ±45 degrees, the normal operating range for the human ankle.

[**0032**] During controlled plantar flexion (CP), normal ankle function can be modeled as a linear rotational spring where ankle moment is proportional to ankle position. Thus, during the CP phase of walking, a linear torsional spring control can be used for the orthotic ankle joint. As a criterion for selecting a desired stiffness of the orthotic torsional spring, the controller can be used to analyze the ground reaction force generated at the moment of forefoot impact after each walking step. The extent of foot slap can be deemed too extreme, and the CP stiffness too low, if a high frequency force spike occurs at the moment of forefoot collision.

[**0033**] In **FIGS. 4A and 4B**, a representative forefoot force signal from a drop foot participant is compared to a forefoot force signal from a normal participant. Both participants wore the AAFO **10** under a zero impedance control, and the forefoot force signal was computed from the sum of all four force transducer signals measured in the forefoot region. In **FIG. 4A**, a dual peak force pattern indicates the occurrence of foot slap in the drop foot participant, whereas in **FIG. 4B**, the lack of a dual force spike indicates that no foot slap had occurred in the normal participant.

[**0034**] To detect the dual peaks and the occurrence of foot slap, the AAFO controller can numerically differentiate the forefoot force and then filter that signal using a second order Butterworth filter with a cutoff frequency of about 0.6 Hz. If substantial foot slap occurs, the differential of the forefoot force is negative, and the stiffness of the orthotic torsional spring stiffness can be incremented. The CP stiffness can be started at zero and incremented by the rules shown in Table I, where the incremental stiffness ($\Delta\Gamma$) was 5.7 Nm/rad (0.1 Nm/deg), approximately 2% of the anticipated final ankle stiffness.

TABLE I

Number of slaps in last 5 steps (n)	Change in Ankle Stiffness
0	$-\Delta\Gamma$
1	0
2–5	$(n - 1)\,\Delta\Gamma$

[**0035**] Gait speed is an important step-to-step gait variation for which the AAFO **10** can respond and adapt. In a particular embodiment, the time of foot contact, defined as the time that a foot remains in contact with the ground from heel strike to toe-off, can be used as a measure of forward speed. With an expectation that orthotic CP stiffness should change with gait speed, the full range of gait contact times can be divided into bins, denoting velocity ranges. During each swing phase, stance time can be estimated from the orthotic force transducers **16**, and the participant's time of contact bin, or forward speed range, can be selected. Within each bin, the AAFO controller can optimize the orthotic CP stiffness. In one embodiment, only three bins are necessary to span the full speed range of the participants.

[**0036**] Drop foot participants typically do not experience any difficulties during powered plantar flexion. Hence, the control objective of Contact **2** is to minimize orthotic joint impedance so as not to impede the participants' power plantar flexion movements. During this state, the SEA's **12** target force can be set to zero.

[**0037**] During the swing phase, a second-order, underdamped mechanical model (spring-damper PD control), previously used to characterize normal ankle function, can be used to control the orthotic ankle joint. Using the AAFO **10**, each drop foot participant can walk at slow, self-selected, and fast speeds, and the swing phase ankle angle can be collected on both the affected and unaffected sides. At each speed, orthotic joint stiffness can be increased manually until the early swing phase dorsiflexion velocity measured on the affected side matched the unaffected side. Orthotic joint damping can be increased from zero until unwanted joint oscillations are removed. The final values of stiffness and damping in this particular embodiment are listed in Table II below.

TABLE II

Gait Speed	K (Nm/rad)	B (Nms/rad)
Slow	28.65	0.57
Normal	37.24	1.03
Fast	45.84	1.15

[**0038**] The stiffness and damping values for the drop foot users are not correlated with gait speed directly, but with ranges of stance time, in the same manner to the CP stiffness control described earlier.

EXAMPLE

[**0039**] A clinical evaluation of the AAFO **10** was conducted in the Gait Laboratory at Spaulding Rehabilitation Hospital, Boston, Mass. Drop foot participants having only a unilateral drop foot condition were selected, and on their affected side, participants did not suffer from a gait disability